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REVIEW OF RECENT RESULTS USING COMPUTATIONAL FLUID DYNAMICS SIMULATIONS IN PATIENTS RECEIVING MECHANICAL ASSIST DEVICES FOR END-STAGE HEART FAILURE

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Abstract

Many end-stage heart failure patients are not eligible to undergo heart transplantation due to organ shortage, and even those under consideration for transplantation might suffer long waiting periods. A better understanding of the hemodynamic impact of left ventricular assist devices (LVAD) on the cardiovascular system is therefore of great interest. Computational fluid dynamics (CFD) simulations give the opportunity to study the hemodynamics in this patient population using clinical imaging data such as computed tomographic angiography. This article reviews a recent study series involving patients with pulsatile and constant-flow LVAD devices in which CFD simulations were used to qualitatively and quantitatively assess blood flow dynamics in the thoracic aorta, demonstrating its potential to enhance the information available from medical imaging.

Introduction

The number of patients receiving left ventricular mechanical assist devices (LVAD) is constantly increasing. The concept of long-term mechanical support is gaining more acceptance as a bridge to transplant, bridge to decision and bridge to destination therapy. As these devices may profoundly alter the hemodynamics, especially if they provide constant flow, a need arises for a better understanding of these alterations and the consequent response of the cardiovascular system. These responses may include short-term postoperative complications after implantations and/or long-term changes such as the formation of arteriovenous malformations. Also, elevated wall shear stress (WSS) may mobilize atherosclerotic plaques, leading to thromboembolic complications with ischemic events in the end organs. Another well-known complication in this patient population is the development of aortic insufficiency.

While measurements of hemodynamics in general can be realized using magnetic resonance imaging (MRI), LVAD devices are not MRI-compatible and therefore other means have to be explored to gain insights into blood flow dynamics. An alternative is the use of computational simulations, and here computational fluid dynamics (CFD) offers the opportunity to simulate hemodynamic pressures and flow patterns in LVAD patients. CFD has previously been successful in patients with a variety of cardiovascular diseases, such as cerebral aneurysms, abdominal aneurysms or aortic dissections.[1-6]

Three recent studies applied CFD in LVAD patients to simulate aortic hemodynamic factors, in particular dynamic pressure, WSS, velocity magnitudes, and turbulent dissipation.

In the first study, flow patterns in two of the most common geometries of LVAD cannula insertions were analyzed.[7] In the second study, patients with continuous-flow and pulsatile-flow devices were compared.[8] In the third study analyzing a larger LVAD patient population, CFD was used to assess adverse hemodynamic conditions and correlate the parameters with aortic insufficiency and ischemic events.[9]

The following provides an overview of this study series and explores how CFD simulations are enhancing the information available from medical imaging and may become an integral part of clinical imaging research tailored towards individualized patient management.

Technical Realization of CFD in the Aorta after LVAD Implantation

CFD offers the possibility of simulating flow conditions and deriving quantitative values of hemodynamic parameters. The integration of CFD into the clinical workflow involves close interaction between clinical experts (such as radiologists and surgical specialists) and experts within the biomedical and computer science communities, with essential exchanges of knowledge and feedback from one group to the other (Figure 1).

In our study series, the technical procedure was conducted as follows. Existing DICOM images of clinical computed tomographic angiography (CTA) examinations were exported to a dedicated image processing workstation. Thereafter, images were imported in a processing software (ImageJ, National Institutes of Health, Bethesda, MD). Grey-scale boundaries were enhanced into facilitate further segmentation. Image noise and inhomogeneities were reduced by applying a fast Fourier transform spatial band pass filter. The resulting images were inverted to obtain dark boundaries around the ascending aorta and the lumen of

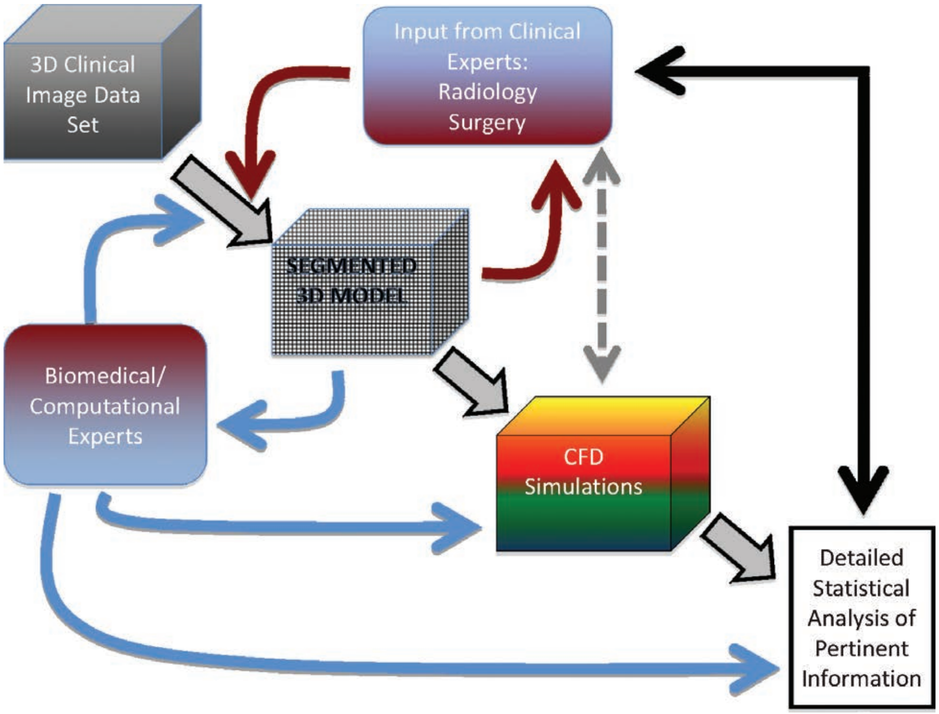

Figure 1. Workflow of a computational fluid dynamics project based on clinical image data, i.e., a retrospective approach. The 3-dimensional dataset (computed tomography images of patients with left ventricular assist devices) are converted into a computational mesh after segmenting the region of interest (gray arrow). At this step, input from the clinical specialities (radiologists, surgeons, cardiologists) is advantageous to the biomedical or computer expert to ensure that the correct model is created. The interaction between the different groups is usually well established as both parties communicate to initiate the project. After this initial interaction, the biomedical group might then continue to conduct the simulation and the analysis. However, from the experience of the authors, another feedback loop at this stage is beneficial to exclude unrealistic assumptions during the simulation and to discuss the validity of first simulation results. The analysis outcome should be a detailed statistical analysis of the simulations, and a report should be created as deliverables to the clinical experts. While it is tempting to provide the appealing color-coded images of the simulation results, a statistical postsimulation report is essential for the quantification of results and trends in the datasets.

the LVAD-outflow cannula. Using 3-dimensional (3D) image processing software (Paraview, Kitware, Inc., Clifton Park, NY), technicians created a 3D reconstruction of the aorta including the LVAD cannula and the surrounding tissue.

The simulation process began with the creation of tetrahedral meshes for the 3D reconstructed aorta, including the LVAD outflow cannula. To reduce computational bandwidth, tetrahedral elements were merged to form polyhedral structures. Boundary conditions for the simulation included constant flow velocity for continuous-flow devices of 1 m/s and a maximum velocity of 0.8 m/s during systole for pulsatile-flow devices, zero pressure at the supra-aortic vessels and the distal aorta, and negligible flow contribution over the aortic valve as verified by echocardiography measurements. Flow conditions were simulated using a turbulent CFD model, the realizable κ-ε model. This model is a two-transport equation (κ equation and ε equation) accounting for turbulent conditions and high velocities developing at the ascending aorta and the cannulation site. It provides an acceptable computational solution for simulating complex and large geometries. Standard wall functions were used, and a laminar stress-strain relationship was assumed. Further assumptions included rigid walls and Newtonian behaviour of blood.[7,8] Results from CFD simulations for LVAD patients can include a variety of hemodynamic parameters such as blood velocity, WSS, and pressure, and the presentation of the results may vary (Figure 2).

Comparison Between Lateral and Anterior Insertion of the Outflow Cannula

In the initial study, two patients with the HeartMate II Left Ventricular Assist System (Thoratec Corporation, Pleasanton, CA) were compared with regard to cannula anastomosis site geometry and its impact on hemodynamic conditions in the ascending aorta.[7] Patient A had lateral geometry of cannula insertion, whereas patient B had anterior insertion of the LVAD outflow cannula. In both patients, streamline analysis showed no significant differences during systole and diastole. Laminar ordered flow was found in the device outflow cannula and in the descending aorta. Flow disturbance and turbulences were highest around the LVAD anastomosis site. In Patient A, this region extended beyond the supra-aortic vessels distally into the aortic arch. In Patient B, unordered flow was restricted to the ascending aorta. In Patient A, the resulting outflow acceleration, with velocities exceeding 1.5 m/sec from unordered flow, impinged on the contralateral aortic wall. This impingement caused an increase of dynamic pressure on the aortic wall reaching values greater than 900 Pa. In Patient B, no points of focally increased dynamic pressures were noted; rather, the highest pressures were noted at the wall of the LVAD cannula, probably because of its tortuous course. The average WSS was approximately 20 times higher in Patient A compared to Patient B. Both patients showed retrograde flow towards the aortic root. However, lower velocities were recorded in Patient B.[7] This initial study pointed out the value of CFD in analysing different geometries, which may have a significant influence on the hemodynamics of the ascending aorta. The results suggest that an anterior geometry when implanting an LVAD may be preferred as it leads to an ordered and thus more benign flow pattern.[7]

Comparing Pulsatile Versus Continuous-Flow LVAD Systems

In the second study, the hemodynamic conditions in the ascending aorta were compared between pulsatile and continuous-flow LVADs. Three patients had the HeartMate II, representing the nonpulsatile assist system, and three patients received the Excor Ventricular Assist System (Berlin Heart Inc., The Woodlands, TX), representing the (biventricular) pulsatile assist system.[8] The pulsatile assist devices showed a minimal variability in the angles between the insertion site of the outflow cannula and the ascending aorta. The angles were as follows: 85°, 95°, and 89°, respectively. The angles varied significantly in the HeartMate II patients: 78°, 41° and 29°. When simulating flow patterns, a retrograde flow towards the aortic root was seen in both device types. However, since continuous-flow devices were routinely implanted more distally in the ascending aorta, larger recirculation patterns with higher vortices were developed in this group,

Figure 2. On top is the segmented model of the left ventricular assist device (LVAD) outflow graft anastomosis site at the thoracic aorta. Below is a color-coded display of three hemodynamic parameters in different presentations. On the left, velocity is shown as a semitransparent volume display (higher velocities are red and more opaque). The center depicts wall shear stress (WSS) at the points of the computational grid; and the right shows total pressure in a 3-dimensional surface display. For the velocity, a semitransparent display is advantageous as the velocity jet inside the computational model is mostly of interest (velocity is zero by definition at the wall). For WSS, small variation may be important (focal maxima of shear or force may be indicative of future problems), therefore a granular display is important. Total pressure mainly illustrates the overall general pressure situation (here, blue is high pressure due to the inflow boundaries of the LVAD graft and the aortic valve). No details along the computational model are usually needed.

causing more dynamic pressure on the aortic wall. In contrast, only one patient with a pulsatile-flow device developed high turbulent retrograde flow with increased pressure on the contralateral aortic wall. Streamlines, visualized in patients with pulsatile devices, appeared laminar and generally ordered downstream from the insertion site. Patients with continuous-flow devices showed unordered flow in the aortic arch, and ordered laminar flow seemed to start more distally in the descending aorta. Wall shear stress on the aortic wall was especially increased on the opposite site of the cannula insertion site with subsequent impingement of the aortic wall.[8] The results of this study indicate a favorable hemodynamic pattern in pulsatile LVADs that is in accordance with clinical considerations.[8]

Correlation of Adverse Effects Using CFD in LVAD Patients with Hemodynamic Alterations in the Ascending Aorta

In a larger LVAD patient population, hemodynamic alterations in the thoracic aorta were correlated for the first time with actual clinical events.[9] Postimplantation, three patients developed aortic insufficiency (AI) diagnosed on transthoracic echocardiography while two patients had ischemic events (ischemic stroke and colon ischemia). LVAD outflow graft orientation was defined by two angles, the azimuth angle measured between the aortic arch and the outflow graft, and the polar angle measured between the anterior wall of the ascending aorta and the outflow graft. Hemodynamics was described in relationship to the cannula orientation. Applying a linear discriminant analysis, the relationship between cannula orientation (azimuth and polar angle) and the resulting hemodynamic alterations (pressure gradients, WSS, and turbulent energy dissipation) was correlated to the adverse clinical events. Significant variations dependent on azimuth and polar angle were found. Dynamic pressures were increased in close proximity of the entrance site of the LVAD outflow cannula. In two patients, the high-pressure zone with subsequent impingement was projected on the contralateral aortic wall. Two other patients showed high focal pressure zones on the ipsilateral aortic wall distal to the anastomosis site. One patient showed no high focal pressure zones at all. In all five cases, unordered flow was noted in the ascending aorta and the aortic arch—most markedly in the ischemic stroke and two AI cases. In one of the two AI cases, unordered flow extended to the descending aorta. The orientation of the LVAD cannula described by the azimuth and polar angle correlated with pressure zones, WSS, and dissipation of turbulent energy. Especially noteworthy is the statistically significant elevated velocity magnitude of retrograde flow in the ascending aorta of patients who developed AI.[9] The data from this patient series shows that LVAD implantation leads to altered aortic hemodynamics. These alterations may at least partially explain postoperative complications such as AI or ischemic events.[9]

Discussion

The initial study described above demonstrated that in LVAD patients receiving continuous-flow assist devices, an anterior geometry of the cannula anastomosis showed less unordered flow at the root of the supra-aortic vessels and no impingement pressure zones at the contralateral aortic wall, thus reducing the amount of retrograde flow and pressure on the aortic root. To achieve the correct position in the ascending aorta, the pump implantation should be done via median sternotomy as opposed to a lateral minithoracotomy, in which the outflow graft is attached to the descending aorta. The inflow cannula is attached in its usual fashion after initiation of cardiopulmonary bypass, while the outflow cannula is anastomosed using a side-biting clamp to the ascending aorta.[10] Care must be taken while directing the outflow graft to avoid kinking or torque on the device.[11] The anastomosis site should be inspected for calcification, and the selection of cannula location should take into consideration the possibility of redo surgery for cardiac transplantation.[12]

Furthermore, pulsatile flow showed more favorable hemodynamic conditions including reduced WSS and lower pressures in the ascending aorta when compared to continuous-flow LVADs. Nonetheless, the number of nonpulsatile-flow devices has been increasing in clinical applications since those devices are usually smaller and produce less frequent bleeding and infections.[13] However, the loss of pulsatility is unphysiologic and interferes with the integrity of the circulatory system. Different studies show continuous flow to have a substantial impact on the aortic and arterial walls, considering the thinning of the medial layer and volume ratio of smooth muscle cells.[14] Cyclic stretch of smooth muscle cells is a main cause of proliferation and is directly responsible for the production of autocrine growth factors.[15,16] Hence the loss of these stimulatory factors in nonpulsatile devices results in reduced smooth muscle contractility and elasticity due to an increased ratio of elastin to collagen. Furthermore, continuous-flow devices lead to a reduction in pulse pressure that in turn increases systemic vascular resistance.[17] Eventually this interference with the circulatory system's mechanical properties may lead to an increased afterload and impact the regulatory properties of blood flow to end organs.[14,18] In a retrospective study analyzing the possibility of weaning after left ventricular assist therapy, pulsatile devices showed a three-fold higher recovery rate when compared to continuous-flow devices.[19] This might be attributed to better unloading of the left ventricle when using pulsatile devices[20,21] and decreased coronary blood flow in continuous-flow devices.[22] Furthermore, continuous unloading of the left ventricle has been shown to negatively impact the pressure volume relationship of the heart and deranges vascular and hemodynamic energy utilization.[23]

One of the leading complications after LVAD implantation is the development and progression of aortic valve disease, especially aortic regurgitation.[24-27] Severe aortic insufficiency can lead to heart failure symptoms and may require surgical intervention to restore cardiac function.[28] The use of continuous-flow devices has been reported as an independent risk factor for developing aortic insufficiency.[29] High velocity of retrograde flow towards the aortic root might be partly responsible for this phenomenon. Other contributors to aortic valve malcoaptation may be changes in aortic wall elasticity and high diastolic intraluminal pressures. In addition, retrograde flow in the ascending aorta causes early valve closure and shortened systole.[22,30]

Severe aortic insufficiency can lead to LVAD failure since the blood recirculates through the incompetent aortic valve to the LVAD inflow cannula, impeding myocardial recovery due to failed unloading of the left ventricle; it can also lead to end-organ malperfusion since the blood is short-circuited.[30] Furthermore, severe aortic insufficiency increases the risk of thrombosis and blood hemolysis.[31-33]

Conclusion

These studies demonstrated that computational fluid dynamics simulations enhance the information in clinical image data and may have an application in clinical investigations of hemodynamic alterations. The results also suggest how clinicians can adapt the location of the anastomosis site to optimize flow and avoid both turbulences and increased WSS and pressure on the aortic root. The remaining challenge is to integrate these simulations into the clinical workflow so that this technology may be used routinely.

Conflict of Interest Disclosure: The authors have completed and submitted the *Methodist DeBakey Cardiovascular Journal* Conflict of Interest Statement and none were reported.
Funding/Support: The authors have no funding disclosures.
Keywords: computational fluid dynamics, left ventricular assist device, heart failure

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